

Locomotor Training: The Effects of Treadmill Speed and Body Weight Support on Gait Kinematics

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Abstract

Body Weight Supported Treadmill Training (BWSTT) is a rehabilitation method which can help individuals with incomplete Spinal Cord Injuries (SCIs) regain the ability to walk. In BWSTT, a harness is used to support part of the patient's body weight above a treadmill while therapists assist the patient in performing stepping motions. Following SCI, recovery of function is achieved by reorganization of neural pathways in the spinal cord, which is dependent on task-specific rehabilitation. BWSTT is a task-specific program which focuses on replicating the forces and motions experienced during normal walking to help patients achieve recovery of walking ability. While BWSTT has shown positive results, there is no quantitative evidence for which training conditions, specifically treadmill speed and amount of body weight support, most effectively target the muscles to restore normal gait. This study will determine how varying treadmill speed and percentage of body weight support affects the kinematic gait parameters of normal walking.

We collected motion capture data from 4 healthy subjects (3 male, 1 female), as they walked on an instrumented treadmill with varied body weight support (0,30,50, and 70% of weight supported) and treadmill speed (self-selected speed (SS), 0.5xSS, and 1.5xSS). We analyzed the motion capture data and developed dynamic, patient specific models using an open-

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source software for dynamic simulations of human movement. Inverse kinematics techniques were used to calculate joint angles and spatiotemporal gait parameters.

Preliminary results suggest that high levels of body weight support cause large differences in lower extremity joint angles, particularly at the ankle. Percentage of stance during the gait cycle appears to decrease with increased body weight support, and also with increased speed. These data suggest that walking with very high levels of body weight support may not replicate normal walking conditions closely enough to produce the most effective rehabilitation strategy.

The results of this study will shed insight into which BWSTT conditions cause gait kinematics to deviate from those seen in normal walking patterns. Determining the affect of specific training conditions will help therapists make more informed decisions when setting treadmill speed and body weight support levels, which could lead to the development of more efficient rehabilitation programs.

Introduction

Spinal Cord Injury

More than 250,000 people in the United States suffer from debilitating spinal cord injuries (SCIs) and nearly 12,000 new injuries occur each year (UAB Facts and Stats). These injuries are most commonly the result of traumatic situations such as automobile accidents (42%), falls (21%) or violence (15%) (UAB, 2006). Since 2005 the average age at injury is 39.5 years, and males are most commonly affected (77.8%) (UAB Facts and Stats).

An injury is considered a SCI when damage to the spinal cord results in the loss of feeling or mobility below the level of injury. All SCIs are classified as complete or incomplete injuries (Trieschmann, 1988). Sensory and motor functions throughout the body are controlled

by nerves extending from the spinal cord at different levels (Trieschmann, 1988). A SCI will affect all functions controlled by spinal nerves below the location of injury (Trieschmann, 1988). For example, any injury above the L5 vertebrae will affect the leg muscles and will inhibit walking ability (figure 1).

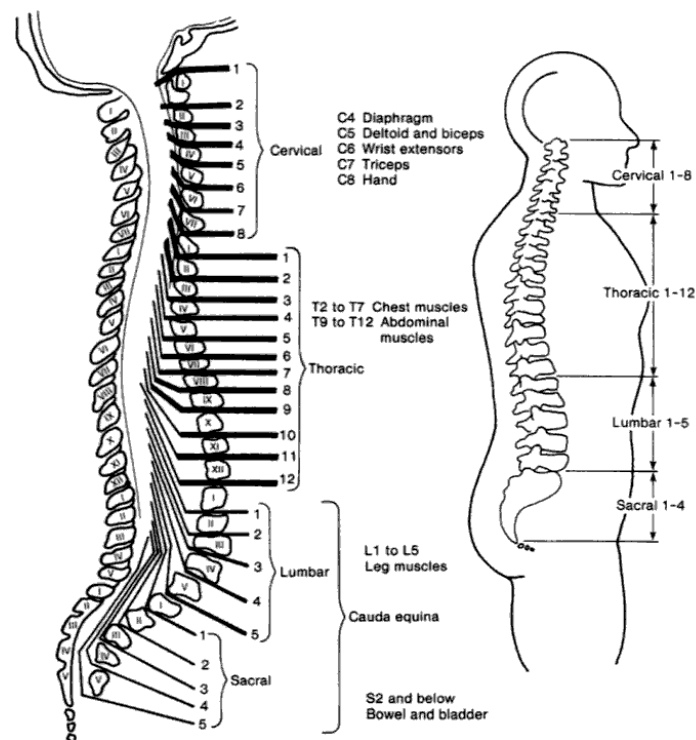


Figure 1 Spinal nerves extending from the spinal cord control different functions throughout the body (Trieschmann, 1988).

Traditional Rehabilitation vs. Body Weight Supported Treadmill Training

Until recently, it had long been believed that patients with SCIs would not be able to recover pre-injury function as a result of physical rehabilitation (Behrman and Harkema, 2007). Physical therapy programs were designed to teach patients with SCIs to compensate for the loss of motor control by using new strategies to accomplish everyday tasks (Behrman and Harkema, 2007). Traditional rehabilitation focuses on muscle and neurological training, but does not target the neurons in the spinal cord which are believed to control walking patterns.

In contrast, Body Weight Supported Treadmill Training (BWSTT) for SCI rehabilitation is an activity-based therapy which aims to help patients recover neural control of locomotion. During BWSTT, a patient is supported by a harness suspended above a treadmill while trained therapists provide manual assistance as necessary to help the patient perform stepping motions (figure 2). With therapist assistance the patient can experience the motions and the forces associated with normal walking, which helps the spinal cord redevelop control of repetitive stepping motions (Van de Crommert et al., 1998).



Figure 2 Patient in BWSTT environment. A harness supports part of the patient's body weight while therapists assist the motion of his feet and provide support at his hips. Photo courtesy of MediaSource, Dublin, OH

In BWSTT, patients no longer have to accomplish tasks of increasing difficulty one at a time, and instead can train in an environment that encourages simultaneous development in posture, balance, and stepping which are the three most important components of locomotion (Barbeau et al., 1987).

BWSTT has helped patients with neurological disorders, such as SCI or stroke, to redevelop the ability to walk over ground and to reduce their need for assistive devices. Wernig and colleagues (Wernig et al., 1995) studied the effectiveness of BWSTT in 89 patients with SCI and observed 76% of the patients learned how to walk independently (Wernig et al., 1995). Patients maintain the benefits achieved during rehabilitation even years after the completion of BWSTT programs (Wernig et al., 1998).

However, although the overall results of BWSTT have been positive, the gait patterns resulting from this rehabilitation program are less than ideal (Finch et al., 1991). Patients who have achieved community ambulation often struggle to control trunk movement and lack proper ankle motion (Finch et al., 1991). These results might suggest that current BWSTT programs may not be efficiently facilitating the development of normal walking patterns as efficiently as they could be.

A major challenge associated with BWSTT is determining appropriate parameters such as treadmill speed and amount of body weight support. These parameters are currently set by a physical therapist's perception of "what looks right" while the patient is on the treadmill. This subjective method of program design is highly dependent on the experience of the therapist. Training parameters that are believed to facilitate locomotor patterns may not actually be targeting the appropriate muscles or neural networks (Hidler, 2005). In order to improve the functional results of BWSTT programs, it is important to quantitatively determine how different training conditions could affect walking performance (Hidler, 2005).

Research performed using healthy subjects in the BWSTT environment has determined that increasing body weight support has a statistically significant effect on temporal patterns, joint angles, and muscle activation patterns and amplitudes (Finch et al., 1991; Threlkeld et al.,

2003). Temporal phases associated with balance, such as the percentage of stance and total double limb support time, decrease with increasing amounts of BWS (Finch et al., 1991; Threlkeld et al., 2003). Joint angles observed during walking with high levels of BWS (50-70%) differ in magnitude and shape from joint angles observed during normal walking. At high levels of BWS subjects allow the treadmill to pull the foot into extended plantar flexion during late stance, resulting in abnormal ankle, knee and hip angles (Threlkeld et al., 2003).

While these and other studies have observed the effects of body weight support for constant treadmill speeds, this study aims to determine how changing both treadmill speed and body weight support affect the kinematic and temporal gait patterns. And also to determine whether changes in knee flexion angle, ankle plantar flexion or dorsiflexion angle, and percentage of stance are more affected by treadmill speed or by the application of body weight support

Methods

Four subjects (3 male, 1 female) participated in this study. Subjects ranged between 23 and 30 years of age with a mean age of 25.25 years \pm 3.304 (S.D.). By self-report, all subjects were free of neurological disorders, back, leg, and foot pain, and any injuries which could interfere with normal walking patterns. The procedures of the study were explained and informed consent was obtained from all subjects.

Prior to beginning the experimental protocol, subjects were assisted into the body weight support harness and performed a ten meter over ground walk to determine their self-selected walking speed. A medical scale was used to determine starting body weight which was used for calculating the amount of support to be supplied. Reflective markers were placed on the subject's skin using a modified Point-Cluster Technique (Andriacchi et al., 1998). An over-

abundance of markers were placed on the thigh and shank segments to define cluster coordinate systems tied to anatomically relevant coordinate systems calculated with the subject at rest (Andriacchi et al., 1998).

Subjects walked on a split belt instrumented treadmill (Bertec Corp., Columbus, OH) over a range of treadmill speeds with various levels of body weight support applied. Ground reaction forces were collected at 2,000 Hz and low-pass filtered with a fifth order Butterworth filter (10 Hz cutoff frequency). Body weight support was provided by a medical body weight support harness and a closed-loop pneumatic force control system. The system used a pneumatic cylinder with pressure control and a feedback loop with a continuous signal from a force transducer in series with the support cable in order to supply a steady amount of support to the subject via the harness (Vigor Equipment, Stevensville, MI; Tescom, Elk River, MN). Kinematic gait data were collected using a seven camera VICON motion analysis system (Vicon Mxcameras, Vicon, Inc.).

Subjects performed twelve thirty-second walking trials on the treadmill for four levels of body weight support (0%, 30%, 50% and 70% of weight supported) and three treadmill speeds (Self-Selected (SS), 0.5 x SS, and 1.5 x SS). Conditions were randomized for each subject.

Kinematic data were processed in the Vicon Nexus environment and sent to an open source biomechanical modeling program for further analysis (OpenSim, Stanford University, Stanford CA). Dynamic musculoskeletal models were used in combination with experimental data to create three-dimensional simulations of each subject's walking trials. We used the inverse kinematics tool in OpenSim, which uses a least squares approach to minimize the difference between the experimental marker location and the virtual markers on the model while maintaining joint constraints, to determine the joint angles that best reproduce the mental

marker data (Delp et al., 2007). We used inverse kinematics to determine knee flexion angles, and ankle plantar flexion and dorsi flexion angles throughout the gait cycle.

The knee flexion angles, ankle angles, and percentages of stance time were used in a 3 x 4 repeated-measures analysis of variance (ANOVA). We used this statistical test to determine if body weight support and treadmill speed are factors that affect the kinematics of normal walking. We examined the assumption of sphericity with Mauchly's Test of Sphericity. If the assumption of sphericity was violated, we used the Greenhouse-Geisser correction factor. We performed post-hoc tests using a Bonferroni adjustment for multiple comparisons for factors that were significantly changed at the $p < 0.05$ level. Statistical tests were calculated with a statistical software package (SPSS Inc, Chicago, IL).

Results

Percentage of Stance

Mean percentage of stance for the baseline condition (0% BWS, SS speed) was $61.9068\% \pm 1.1394$. Increasing body weight support caused the percentage of stance to decrease; for the 70% BWS, SS speed condition percentage of stance was $54.5127\% \pm 3.3206$. With BWS held constant, the percentage of stance was found to increase with decreasing treadmill speed, and to decrease with increasing treadmill speed. The combined effects of BWS and treadmill speed on the percentage of stance can be seen in Figure 3a. Percentage of stance decreases both as a result of increased BWS and increased speed.

Knee Flexion Angle

At 0% BWS and SS speed the mean maximum knee flexion angles were $65.8817^\circ \pm 7.6752$. A decrease in maximum knee flexion angle was found when body weight support was increased; the mean knee flexion angle for the 70% BWS, SS speed case was $57.1544^\circ \pm 5.1002$.

Increasing treadmill speed from 0.5 x SS to 1.5 x SS while holding BWS constant generally showed an increase in knee flexion angle. Highest knee flexion angles were seen at the 0% BWS, 1.5 x SS condition ($67.2254^{\circ} \pm 11.7902$) and lowest maximum knee flexion angles were seen at the 70% BWS, 0.5 x SS condition ($54.7185^{\circ} \pm 7.7344$). Figure 3b shows the combined effects of speed and BWS on the maximum knee flexion angle; large differences are seen

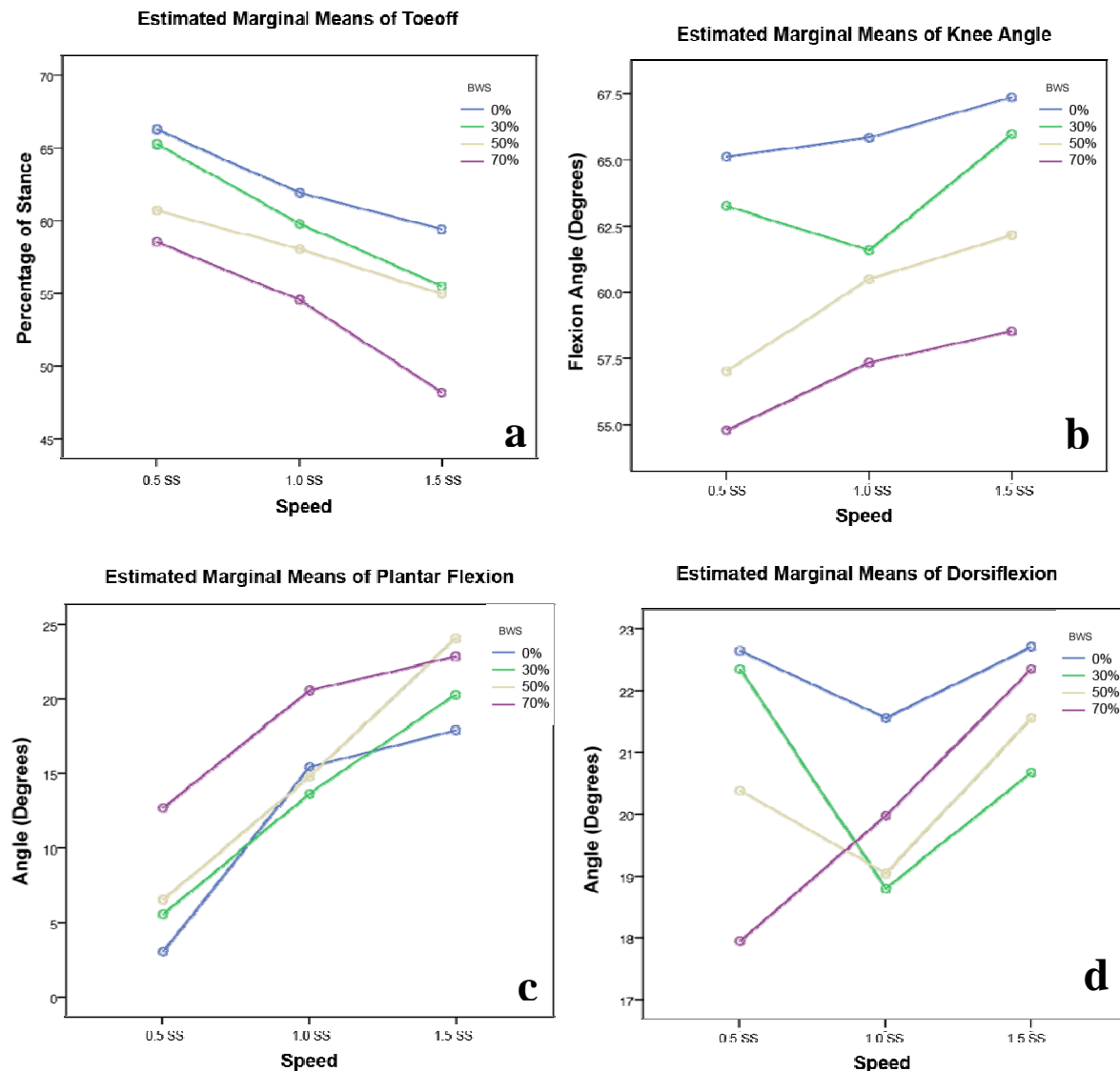


Figure 3 Plots of estimated marginal means of joint angles and temporal patterns for changing speed and body weight support (BWS). The four levels of body weight support (0%, 30%, 50%, and 70%) are shown as separate lines on each plot. Percentage of stance (a), maximum knee

flexion angle (b), maximum ankle plantar flexion angle (c), and maximum ankle dorsiflexion angle (d) are shown here.

between the BWS conditions and a general trend toward increasing flexion angle appears with increased speed.

Ankle Plantar Flexion

Mean maximum ankle plantar flexion angles for the 0% BWS, SS condition were $15.2581^{\circ} \pm 4.3075$. As treadmill speed increased from 0.5 x SS to 1.5 x SS for a constant level of BWS the maximum plantar flexion angle also increased. As BWS increased the plantar flexion angle showed a general trend of increasing as well. Highest plantar flexion angles were observed at the 50% BWS, 1.5 SS condition ($23.8132^{\circ} \pm 5.4955$) and lowest plantar flexion angles were observed at the 0% BWS, 0.5 x SS condition ($2.8329^{\circ} \pm 1.2953$). Figure 3c shows the combined effects of treadmill speed and BWS on the maximum ankle plantar flexion angles.

Ankle Dorsi Flexion

Mean maximum ankle dorsi flexion angles for the 0% BWS, SS condition were $21.6129^{\circ} \pm 1.5341$. Maximum dorsi flexion angles experienced only small changes with the application of BWS and for changing treadmill speeds. The maximum dorsi flexion angles were observed at the 0% BWS, 0.5 x SS condition ($22.6554^{\circ} \pm 3.2854$) and minimum angles were observed at the 70% BWS, 0.5 x SS condition ($18.0576^{\circ} \pm 1.5242$). Figure 3d shows the changes in dorsi flexion angle as a result of treadmill speed and BWS; no general trends appear to exist based on the current information.

Discussion

BWSTT conditions are currently set by therapists based on what looks right and allows the patient to participate in the exercises. This subjective determination results in the use of a

wide range of treadmill speeds and levels of body weight support during rehabilitation for patients with SCIs. The treadmill speeds and levels of body weight support examined in this study fall within the range of conditions used at rehabilitation centers across the country. The strength of BWSTT rehabilitation for SCI is its ability to provide the patient with the forces and motions associated with walking. This study aims to determine how kinematic and temporal gait patterns are affected by treadmill speed and body weight support. The results of this study may help therapists make more informed decisions when choosing treadmill speed and levels of body weight support during BWSTT rehabilitation for patients with SCIs.

Previous studies have looked at the effects of body weight support on kinematic parameters for specified treadmill speeds. Finch and colleagues studied the effect of various levels of body weight support and found that the percentage of stance and the maximum knee flexion angle decreased progressively with increased body weight support (Finch et al., 1991). We observed a similar decrease in percentage of stance with increased body weight support when treadmill speed was held constant at any speed, and body weight support was altered. Whether the subject walked at 0.5xSS, SS, or 1.5xSS as body weight support increased, the percentage of stance decreased. Similarly, in agreement with the results from Finch et al. we observed that as treadmill speed was held constant at any speed the maximum knee flexion angle decreased with increasing body weight support.

Biomechanical principles suggest that as walking speed increases the percentage of stance decreases. This is evident as walking speed increases until it transitions to running where a very short stance phase is followed by flight time. As expected, we observed a decrease in percentage of stance as treadmill speed increased. This trend was consistent for increasing speed regardless of the amount of body weight support that was provided.

From the plots in Figure 3, it appears that treadmill speed and BWS do not have equal effects on lower extremity kinematics and temporal patterns. Percentage of stance appears to be largely affected by both treadmill speed and percentage of BWS; however, the plantar flexion angle is more greatly affected by increasing treadmill speed than by increasing BWS. Similarly, knee flexion angles experience a greater change due to the application of BWS than they do to changing treadmill speeds. These relationships between changing kinematics and speed or BWS may be important clinically for targeting specific gait parameters during rehabilitation.

A number of possible limitations exist in this study. Potential limitations include the small sample size that has been processed to date; the data presented here represent the results from only four subjects; although data has been collected from twelve subjects the remaining data is still being processed. Another possible limitation of this study is the use of self-selected rather than fixed treadmill speeds. Self-selected speeds were used so that all subjects would be walking at a “slow”, “normal”, and “fast” walking speed, regardless of leg-length. However, in choosing a self-selected speed subjects were encouraged to walk at “a leisurely walking pace” which could be interpreted differently by individual subjects. Choosing fixed treadmill speeds could have eliminated the misinterpretation of self-selected speed, but fixed speeds may result in very different kinematics and kinetics for a subject depending on leg length. Finally, kinematics and kinetics were recorded in a laboratory setting on a split-belt treadmill with the subject wearing a medical support harness and reflective markers for motion capture. These environmental factors and the added equipment may have altered gait patterns during data collection.

This study found that increasing levels of body weight support alter joint kinematics and temporal patterns significantly, regardless of treadmill speed. It is important to determine that

differences in lower extremity kinematics and temporal patterns do exist with changing treadmill speed and BWS. Based on the results of this study, the influence of treadmill speed and body weight support do not appear to have same affect on joint angles and temporal patterns. Further work is required to determine the underlying biomechanical mechanism responsible for altering joint kinematics. This study and the future work to be performed will shed insight into which training parameters could provide patients with kinematics and kinetics that will be advantageous to efficient recovery of over ground walking.

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